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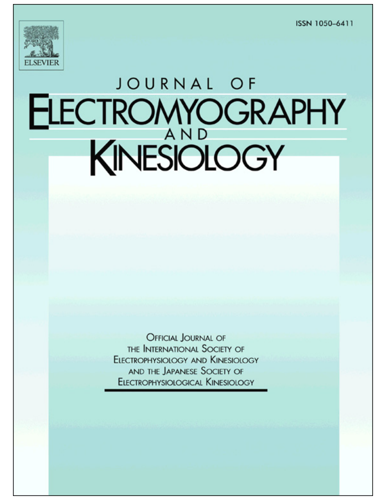

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ARE REGIONS OF THE LUMBAR MULTIFIDUS DIFFERENTIALLY ACTIVATED DURING WALKING AT VARIED SPEED AND INCLINATION?

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ARE REGIONS OF THE LUMBAR MULTIFIDUS DIFFERENTIALLY ACTIVATED DURING WALKING AT VARIED SPEED AND INCLINATION?

ABSTRACT

PURPOSE: Lumbar multifidus is a complex muscle with multi-fascicular morphology shown to be differentially controlled in healthy individuals during sagittal-plane motion. The normal behaviour of multifidus muscle regions during walking has only received modest attention in the literature. This study aimed to determine activation patterns for deep and superficial multifidus in young adults during walking at different speeds and inclination.

METHODS: This observational cohort study evaluated ten healthy volunteers in their twenties (three women, seven men) as they walked on a treadmill in eight conditions; at 2km/h and 4km/h, each at 0, 1, 5, and 10° inclination. Intramuscular EMG was recorded from the deep and superficial multifidus unilaterally at L5. Activity was characterized by: amplitude of the peak of activation, position of peak within the gait cycle (0-100%), and duration relative to the full gait cycle.

RESULTS: Across all conditions superficial multifidus showed higher normalized EMG amplitude ($p<0.01$); superficial multifidus peak amplitude was 232±115% higher when walking at 4km/h/10°, versus only 172±77% higher for deeper region ($p<0.01$). The percentage of the gait cycle where peak EMG amplitude was detected did not differ between regions (49±13%). Deep multifidus duration of activation was longer when walking at the faster vs slower speed at all inclinations ($p<0.01$), which was not evident for superficial multifidus ($p<0.05$). Thus, a significantly longer activation of deep multifidus was observed compared to superficial multifidus when walking at 4km/h ($p<0.05$).

CONCLUSIONS: Differential activation within lumbar multifidus was shown in young adults during walking. The prolonged, more tonic activation of deep relative to superficial regions of multifidus during gait supports a postural function of deeper fibres.

Key Words: Lumbar spine; Multifidus; Intramuscular Electromyography; Gait; Function

INTRODUCTION

Multifidus is a complex muscle in the thoracolumbar region, with differential activation shown in healthy individuals in the sagittal plane (Claus et al., 2009; MacDonald et al., 2009; O'Sullivan et al., 2006), which is likely due to its multi-fascicular morphology. Deepest fibres attaching to the innermost spinous process span one (thoracic) or two (lumbar) motion segments (Cornwall et al., 2011; Macintosh et al., 1986), are closest to the axis of rotation, and are purported to provide support of intervertebral motion (Macintosh and Bogduk, 1986; Macintosh et al., 1986). Relatively superficial fibres of the multifidus span four (lumbar) or more (thoracic) segments (Cornwall et al., 2011, Macintosh et al., 1986) and act as an agonist for trunk and lumbar extension (Macintosh and Bogduk, 1986). Multifidus increases in volume caudally in the lumbar spine and is largest at the level of the fifth lumbar vertebra, while the more lateral erector spinae (longissimus, and iliocostalis) diminish caudally to terminate at the ilium (Cornwall et al., 2011; Crawford et al., 2015). As such, the role of multifidus is crucial at the lowest spinal levels where degenerative change (Brinjikji et al., 2015; Crawford et al., 2015) and injury are most prevalent.

With a parallel architecture and morphology that supports tonic activity (Cornwall et al., 2011), short/deep and long/superficial paravertebral muscle fascicles are differentially activated in healthy volunteers during tasks involving altered spinal posture (Claus et al., 2009; MacDonald et al., 2009; O'Sullivan et al., 2006). Specifically, activity involving an increased lumbar lordosis results in greater and earlier activity of multifidus (medial), with limited impact on iliocostalis (lateral) (Claus et al., 2009; O'Sullivan et al., 2006). Additionally, predictable trunk perturbations are associated with earlier activation of deep versus superficial fascicles of the multifidus (MacDonald et al., 2009; Moseley et al. , 2002). Surprisingly, very few studies have investigated either the thoracolumbar spinal posture or paravertebral muscle activity during gait in asymptomatic adults (Lamoth et al., 2004; Lee et al., 2014; Saunders et al., 2004; Saunders et al., 2005; Thorstensson et al., 1982), or in adults with back pain (Lamoth et al., 2004; Lamoth et al., 2006), despite walking being commonly promoted as beneficial in optimising low back health. An improved understanding of the neuromuscular mechanics of the lumbar spine appears necessary to better inform the use of walking as a therapeutic strategy.

In the only study that we are aware of that examines inclination and speed parameters together during walking, Lee et al. (2014) described differing activity within the lumbar paravertebral muscles of male healthy volunteers. Multifidus showed higher activation at L3 and L5 levels with increasing speed (erector spinae activity did not change), and higher activation of both muscles at L3 with increased inclination. However, their methods used surface EMG and the potential cross-talk from erector spinae at the higher lumbar level limits the confidence of their results in promoting variable inclination for differential activation of the paravertebral muscles. Based on a slightly older and predominantly male (six of seven) cohort to examine postural and respiratory trunk muscle activation during walking and running at variable speeds, Saunders et al. (2004) utilised intramuscular EMG to measure activity of the deep and superficial regions of multifidus at L4. Comparing their two walking speed conditions, biphasic activity was shown for both regions at their slow, and triphasic activity at their faster speed, with different timing noted according to foot strike between the multifidus regions. However, despite kinematic adaptations, the authors reported no differential activity within multifidus during walking, but a trend for increased duration of activation and period of activity with faster locomotion speeds progressing from walking to running. A concentrated evaluation of the influence of modifiable walking parameters on multifidus activity appears warranted.

We therefore aimed to determine the activation of deep and superficial multifidus in young adults when walking on a treadmill under various conditions. We first hypothesised that greater activation would occur in the superficial compared to the deeper region of multifidus secondary to their multi-level mechanical advantage as agonists for trunk mobility, and second, with relatively challenging walking conditions involving higher speed and/or inclination as a response to increasing demand. Third, we expected differential activity to be demonstrated between deep and superficial multifidus regions in relation to the gait cycle, and fourth, for trunk inclination to increase and lumbar lordosis to consequently accommodate the postural changes needed with increased inclination. Our final hypothesis expected lower limb kinematics to be influenced by walking condition, and in particular ankle, knee, and hip motion to increase with more demanding conditions (Saunders et al., 2005).

Normative differential activation of multifidus regions is important to establish as this may lead to future work examining the influence of various factors such as age or pain on the neuromuscular control of the lumbar paravertebral muscles during gait.

MATERIALS AND METHODS

Participants

Ten adult volunteers in their twenties (three women, seven men; aged 26.3±2.5 years) participated after initial screening excluding history of low back pain requiring attention from a health care professional, musculoskeletal injury/disorder, cardiovascular or neurological disorders, diabetes, previous infection following clinical needle insertion, coagulation disorders, medications affecting such, or difficulty with treadmill walking. Recruitment was achieved through local advertisements. The study achieved institutional Ethics Committee approval and complied with the Declaration of Helsinki. All volunteers gave their written informed consent before participation.

Electromyography

With subjects in prone-lying, wire electrodes made of Teflon-coated stainless steel (diameter: 0.1 mm; A-M Systems, Carlsborg, WA) were inserted into the deep and superficial regions of the multifidus muscle via 27-gauge hypodermic needles using ultrasound guidance (Echo Blaster, Telemed; 10-MHz linear transducer) according to an established method (MacDonald, Moseley, 2009). Approximately 3-4 mm of insulation was removed from the tip of the wires to obtain an interference EMG signal. Following skin disinfection (injection swabs: 70% isopropylalkohol, 30x30 mm, Selefatrade, Spånga, Sweden), the needles containing the wire were inserted into the muscle belly and removed immediately to leave the wires in the muscle for the duration of the experiment. Signals were acquired in monopolar mode. Reference electrodes were placed over the right iliac crest and posterior superior iliac spine (PSIS) following skin preparation. EMG data was band pass filtered (8th order zero lag band pass 10-500Hz), sampled at 2048 samples/second, 12-bit A/D converted (EMG-USB2, OT

Bioelettronica, Turin, Italy) and saved on a personal computer HDD (OTBiolab software V.2.05, OT Bioelettronica, Turin, Italy) for further analyses.

Motion capture

Tridimensional tracking of gait cycles was achieved using an 8 camera stereo-photogrammetry system (Oqus 300+, Qualisys Gothenburg, Sweden). Retro-reflective, ball-shaped markers were placed on each subject's skin overlying the following landmarks: seventh cervical spinous process (C7), thoracolumbar junction (~T11; TLJ), peak of the lumbar lordosis (~L3; LUM), sacrum (~S2; SACR), and bilateral PSISs, greater trochanters, fibula heads, lateral malleoli, and fifth metatarsals. Kinematic data was sampled at 256 frames/second (Qualysis Track Manager V. 2.8, Qualisys AB Gothenburg, Sweden) together with one analog channel for synchronization (described below), and stored on a hard disk drive for further analyses.

Procedure

Once the electrodes were in place, participants were given time to familiarise themselves with treadmill walking at a self-selected speed and at different inclinations. An investigator controlled the treadmill settings, explained the procedure, and remained in close proximity to the participant and treadmill. Participants were required to walk continuously for two minutes at 2 km/h and then 4 km/h, and at each of 0, 1, 5, and 10° inclination; all inclinations were undertaken in incremental order at 2 km/h before starting the 4 km/h set at zero inclination; 30-45 seconds rest was given (0km/h-0°) between each test condition. Data was captured for the last 90 seconds of each condition.

Data Analysis

Motion capture and EMG data were synchronized by mirroring the deep multifidus EMG channel on the A/D converter of the motion capture system (digitized at 2048 sample/second, 12bit depth) and offline-computing the time delay as the maximum of the cross-correlation function of the two signals (Gizzi et al. , 2011). To account for differing stride cadences at different speeds for each walking condition, the 41 middle gait cycles (i.e. the middle, previous, and following 20 gait cycles) were retained for further analyses. Analyses were performed through custom-written Matlab scripts (Matlab 2013a, Mathworks Inc, Natick, MA, USA).

Kinematic data was low-pass filtered (10 Hz, 2nd order Butterworth filter). Individual gait cycles were identified as two consecutive heel strikes; briefly: right heel strikes were identified as the local maxima (i.e. the sample where the sign of the first derivative of the signal changed from positive to negative) of the sagittal component of the right ankle marker (Zeni et al., 2008); each gait cycle was time-interpolated to obtain a constant length of 200 samples independently on the time duration of the cycle. Across each gait cycle, ankle, knee, hip, lumbar lordosis (defined by SACR, LUM, and TLJ markers), and trunk inclination (C7-SACR line versus vertical) angulation ranges (sagittal plane), and pelvis rotation (transverse plane) were computed. The pelvis was modelled as a rigid body in the proprietary software, and rotation in the transverse plane (left to right) used for analysis (van den Hoorn et al., 2012). For each angle the average range were computed and retained. Hip, knee and ankle angles were averaged across the right and left side.

The EMG envelope was segmented according to the previously identified heel strikes, and time-interpolated to 200 samples. For each muscle, activation was characterized by: the amplitude of the peak of activation (normalized to maximal activity in the 2km/h-0° condition), the position of the peak within the gait cycle (%), and the duration of the muscle activation computed as the percentage of the gait cycle where the EMG activity exceeded a threshold computed as the average plus three times the standard deviation of a 5% portion of the gait cycle with lowest activation (Di Fabio, 1987). Peak amplitude, peak position and activity duration were computed for each of the 41 gait cycles considered for each condition, and average values were retained for statistical analysis.

Statistical Analysis

Peak EMG amplitude were analysed with two-way analysis of variance (ANOVA) with muscle region (superficial multifidus, deep multifidus) and condition (2km/h-1°, 2km/h-5°, 2km/h-10°, 4km/h-0°, 4km/h-1°, 4km/h-5°, 4km/h-10°) as within-subject factors. The percentage of the gait cycle where peak EMG amplitude was detected, and duration of muscle activation expressed as a percentage of gait cycle, were analysed using three-way ANOVA with muscle region, speed (2km/h, 4km/h), and inclination (1°, 5°, 10°) as within-subject factors. Significant differences revealed by ANOVA were followed by Student-Newman-Keuls (SNK) pair-wise comparisons. Results are reported as mean and SD in the text and standard error (SE) in the figures. Statistical analyses were performed with SPSS Version 22.0 (IBM Corp., Armonk, NY, USA). Statistical significance was set at $p<0.05$.

RESULTS

Electromyography

The normalised peak EMG amplitude was dependent on muscle region ($F=8.73$, $p<0.01$) and condition ($F=4.69$, $p<0.001$) but not the interaction between muscle region and condition ($F=0.63$, $p=0.69$). Across all conditions, superficial multifidus showed higher activation compared to deep multifidus (SNK: $p<0.01$, **Figure 1**). Superficial multifidus' normalised peak amplitude increased by 232.3±115.1% when walking at 4km/h-10° versus 172.2±76.6% for deeper multifidus ($p<0.01$). For both multifidus regions, higher normalised peak EMG amplitude was noted at 4km/h-10° compared to walking at 2km/h at 1°, 5° and 10° of inclination ($p<0.01$). Moreover, higher normalised peak EMG amplitude was noted at 4km/h-5° compared to walking at 2km/h-1°&10° ($p<0.05$), and higher normalised peak EMG amplitude was noted for 4km/h at both 0 and 1° inclination compared to walking at 2km/h-1° ($p<0.05$).

The percentage of the gait cycle where peak EMG amplitude was detected (**Figure 2**) did not differ between muscle regions ($F=0.06$, $p=0.80$) or with speed ($F=1.59$, $p=0.20$). However, the time of peak activity did differ with varying inclinations ($F=3.35$, $p<0.05$). Specifically, the peak EMG amplitude occurred earlier in the gait cycle for both muscle regions when walking at 10° compared to all other inclinations (all SNK: $p<0.05$).

The duration of muscle activation, expressed as a percentage of the gait cycle, is reported in **Figure 3**. The duration of activation was dependent on speed ($F=10.73$, $p<0.01$) and the interaction between muscle region and speed ($F=4.74$, $p<0.05$). In general, the duration of activation of all muscle regions increased when walking at the faster speed (SNK: $p<0.001$), however the interaction revealed that deep multifidus was activated for a longer duration compared to superficial multifidus when walking at 4 km/h (SNK: $p<0.05$). The duration of activation within multifidus did not differ when walking at different inclinations ($F=0.74$, $p=0.52$).

Motion analysis

The average range of lumbar lordosis and trunk inclination did not change across any of the conditions. However, range of pelvis rotation increased at the faster speed ($F=93.34$, $p<0.00001$; SNK: $p<0.0001$) but did not differ with varying inclinations ($F=0.97$, $p=0.40$). Hip range differed with speed ($F=13.46$, $p<0.001$) and inclination ($F=4.74$, $p<0.05$) but was not dependent on the interaction between the two factors. Specifically, hip range increased when walking at 4km/h (SNK: $p<0.0001$) and was larger when walking at 10° of inclination compared to 0, 1 and 5° (all SNK: $p<0.05$). Knee range was also larger at 4km/h than 2km/h ($F=27.78$, $p<0.00001$, SNK: $p<0.001$) but did not differ with the varying inclinations ($F=0.31$, $p=0.81$). Ankle range differed with speed ($F=67.84$, $p<0.00001$) and with inclination ($F=6.47$, $p<0.001$) but was not dependent on the interaction between the two factors. Ankle range increased when walking at 4km/h (SNK: $p<0.0001$) and was larger when walking at 10° of inclination compared to 0, 1 and 5° (all SNK: $p<0.05$).

DISCUSSION

Differential activation of the deep and superficial lumbar multifidus was observed in young pain-free individuals during walking under various speed and inclination conditions. A prolonged and lower amplitude activation of the deep compared to the superficial multifidus, particularly at the faster speeds of this common activity, may support a postural and relatively more tonic function of the deeper fibres. This finding is likely grounded in the complex morphology of the lumbar multifidus wherein fibres have a length-dependent spatial distribution (Cornwall et al., 2011) and consequently different mechanical advantage.

Young paravertebral muscles are shown to be primarily comprised of type 1 fibres (Thorstensson and Carlson, 1987), which supports their tonic role. However, while Cornwall et al. (2011) agree that type 1 fibres predominate in these muscles, their elegant cadaveric dissection study showed the lowest proportion in (deep) fascicles spanning two segments, which seems to contradict our speculation. However, their finding was based on a single elderly case and may reflect declining paravertebral muscle quality as is shown by imaging studies of older spines (Crawford et al., 2015; Fortin et al., 2014; Kjaer et al., 2007) secondary to the degenerative cascade (Haig, 2002). Whether our observations for muscle activation obtained from ten young individuals are generalisable to older adults warrants investigation.

The single-segment innervation purported for multifidus versus the poly-segmental innervation described for erector spinae (Bogduk, 1983), appears counterintuitive to differential neuromotor control within multifidus seen in ours and other studies (Claus et al., 2009; MacDonald et al., 2009; O'Sullivan et al., 2006). However, in their single-subject study of a 49 year old man employing EMG to measure activity after lumbar neurotomy, Wu et al. (2000) purport to demonstrate poly-segmental innervation for multifidus. While their interpretation has been criticised (Haig, 2002), potential for mixed innervation patterns within multifidus may exist and should be investigated further. Alternatively, any differential activation seen within multifidus may not depend on a discreet innervation, but instead be predominantly explained on a mechanical basis.

As outlined earlier, Lee et al. (2014) showed higher multifidus activity at L5 (~165%) at 5&6km/h versus their slow (3km/h) speed. Having measured activity at two levels (L3 & L5), they described no influence of gradient on L5 multifidus activity, but higher activation (~130%) at 15% (as different from degrees of the present study) treadmill inclination compared to 0% at the L3 level ($p=0.012$); this suggests differential activity for lumbar multifidus that may be dependent on spinal level. However, potential for cross-talk from erector spinae using surface electrodes exists, and we cannot be confident their observation was specific to multifidus. As such, our study using intramuscular EMG to record multifidus activity at the lowest lumbar segment during walking might be extended to include multiple thoracolumbar levels to examine activation.

In agreement with our findings, increases in the amplitude and relative duration of superficial abdominal and paravertebral muscles are shown with faster gait speeds (Grillner et al., 1978; Lee et al., 2014; Saunders et al., 2004; Saunders et al., 2005; Thorstensson et al., 1982). However, our results indicating longer duration of the deep compared to superficial multifidus with increased walking demand, and the time of peak activity differing with varying inclinations, appears to conflict in part with the study most closely aligned with ours methodologically (Saunders et al., 2004). Saunders et al. (2004) did not examine the influence of gradient, but showed no differential activity within the paraspinal muscles relating to speed of walking. This disagreement may relate to cohort age and the degenerative decline of muscle quality in the thirties (Saunders et al. cases) compared to the twenties (our cases) (Crawford et al., 2015), the different walking conditions tested between studies, potential muscle activation differences at L4 (Saunders et al.) and L5 (our study), and other variables that may influence spinal health.

Earlier activation of multifidus compared to more lateral erector spinae has been shown during predictable trunk perturbations in healthy volunteers (MacDonald et al., 2009; Moseley et al., 2002). While our study focused on multifidus alone, this result aligns with our findings showing an earlier and more prolonged activity of the deep (and more medial) multifidus using a protocol where walking conditions were declared in advance and therefore anticipated by participants. To add to predictability, and on the basis of optimizing safety, our method only recorded data when accommodation to each new condition had occurred.

Lumbar multifidus volume and fat content increase caudally (Crawford et al., 2015), which probably influences its functional activity. Further, muscle quality associates with skeletal spinal curvature (Meakin et al., 2013; Pezolato et al., 2012) as another mechanical variable that may influence activation. In apparent disagreement with this concept, our study showed no influence of walking condition on the sagittal plane kinematic parameters of trunk inclination and lumbar lordosis, but with increased transverse plane pelvic rotation at higher speed. However, we cannot assume that adaptation of spinal curvature as determined from surface analysis, as we have done, adequately reflects change to skeletal curvature given the relationship between both in the lumbar spine is poor (Crawford et al., 2009).

The kinematic results showing no alteration to trunk inclination or lumbar lordosis with walking conditions were contrary to our expectations; we speculated that our highest gradient would be accommodated by increased forward lean and a compensatory alteration of the lumbar lordosis. In support, previous investigators describe altered muscle activity with subtle changes to the sagittal lumbar curvature in sitting (Claus et al., 2009; O'Sullivan et al., 2006). Our lack of trunk kinematic changes in response to more demanding walking conditions is probably best explained by variability observed within our participants (Table 1), the wide variation reported for lumbar curvature in general (Been and Kalichman, 2014), the suitability of each speed for our young adults, and the sensitivity to measure such subtlety of change. The higher pelvis rotation required at more demanding walking conditions in our participants, and that has been shown by others examining lumbopelvic kinematics during walking (Saunders et al., 2005), may indicate that transverse plane motion of the axial skeleton is where adaptation most critically occurs. Notwithstanding this potential, and in agreement with our results, Lamoth et al. (2004) reported no modification to trunk kinematics (including rotation) after experimentally induced lumbar pain or fear of pain during treadmill walking, despite showing subtle effects on the activity of erector spinae. While our study did not examine this aspect, it may be that the higher activation we observed at greater speed and inclination represents an augmented role of multifidus in attenuating the vibration and torque of ground reaction forces (Jonsson, 1970, Thorstensson et al., 1982).

Lower limb kinematics in our young participants generally agreed with expectations wherein pelvis, hip, and ankle motion increased with higher walking demands. Whether this result is generalisable to other populations requires further investigation in light of gait kinematics differing between young and older healthy adults where progression from an ankle- to hip-centred loci occurs (DeVita and Hortobagyi, 2000).

It has been observed that deeper multifidus is difficult for individuals to engage voluntarily (Hides et al., 1998), which has motivated the use of real time ultrasound in clinical practise as a visual biofeedback method for improving its conscious activation (Van et al., 2006). However, real time ultrasound has limited clinical utility given its impracticality for wide adoption and high cost. As such, our finding showing longer and maintained activation of these fibres when walking at 4km/h and 10° inclination may signal a preferential therapeutic condition for activating deeper multifidus. Additionally, highest amplitudes for both multifidus regions in our study were shown for this walking condition, which may support its integration into rehabilitation and activity programmes for generalised lumbar paravertebral muscle health. Whether beneficial change to muscle quality occurs after such a targeted walking intervention would need to be examined in longitudinal studies that concurrently measure features of muscle quality and function.

While our study offers novelty in using intramuscular EMG to directly and precisely measure multifidus activity in asymptomatic young volunteers while walking, our results must be considered in light of the study's limitations. Pain associated with intramuscular EMG has been shown previously (MacDonald et al., 2009) and may have influenced our detected multifidus activity, yet, we believe our treadmill familiarisation period and delayed data capture after condition accommodation mitigated this. Moreover, none of the subjects reported discomfort from the fine wires apart from the initial pain sensation at needle insertion. Our selected slow 2km/h speed may arguably be too slow, although appropriate for some. An alternative might have been to modify speed according to each individual's self-selected preference, but the need for consistency at our highest inclination made this a more complex choice in terms of participant safety. The limited sample size has not allowed for reporting our results according to each sex, yet evidence points to sex-specific activation patterns of trunk muscles during gait (Anders et al., 2009). While Anders et al. showed that mean and relative amplitudes of trunk muscle activation were not generally dependent on sex, multifidus did show differences between men and women at all examined speeds (2-6km/h).

In conclusion, from the conditions investigated, we present 4km/h at10° inclination as a preferential walking condition to optimally engage multifidus in volunteers in their twenties. Differential activation within lumbar multifidus was shown in varied walking conditions. The sustained, lower amplitude activation of the deep relative to superficial regions of multifidus during gait supports a postural function of the deeper fibres. Further studies are warranted to examine the influence of factors such as age or pain on the differential activation of the multifidus during walking, with potential to better inform personalised management of spinal health.

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FIGURE CAPTIONS

Figure 1: Mean (±SE) of the deep and superficial multifidus normalized peak EMG amplitude (%) during walking at different speeds and inclinations. The EMG amplitude has been normalized to maximal activity in the 2km/h-0° condition.

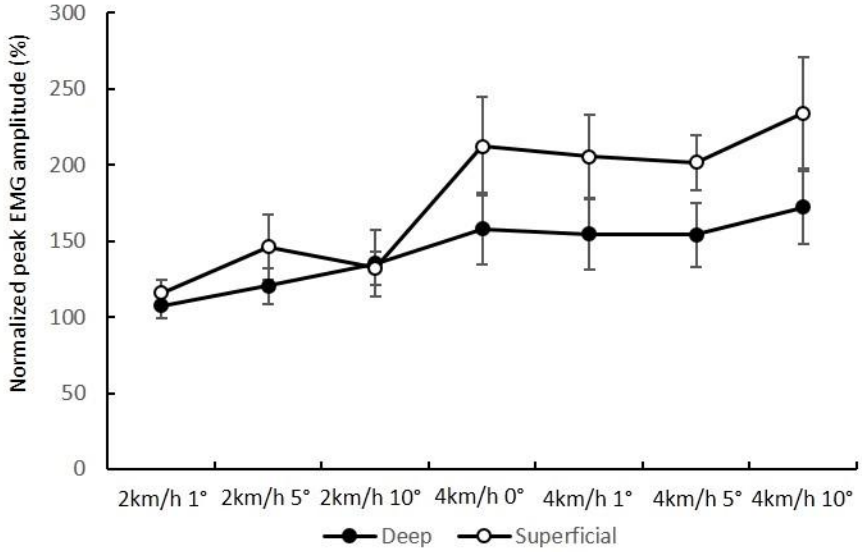
Figure 2: Mean (±SE) of the percent of gait cycle for peak EMG amplitude (%) recorded for the deep and superficial multifidus during walking at different speeds and inclinations.

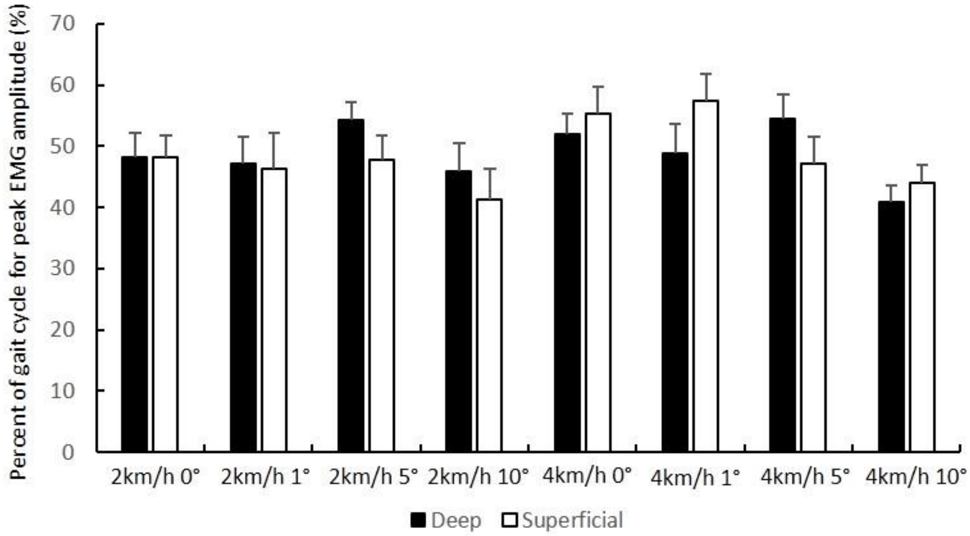
Figure 3: Mean (±SE) of the duration of activation (% gait cycle) recorded for the deep and superficial multifidus during walking at different speeds and inclinations.

TABLE

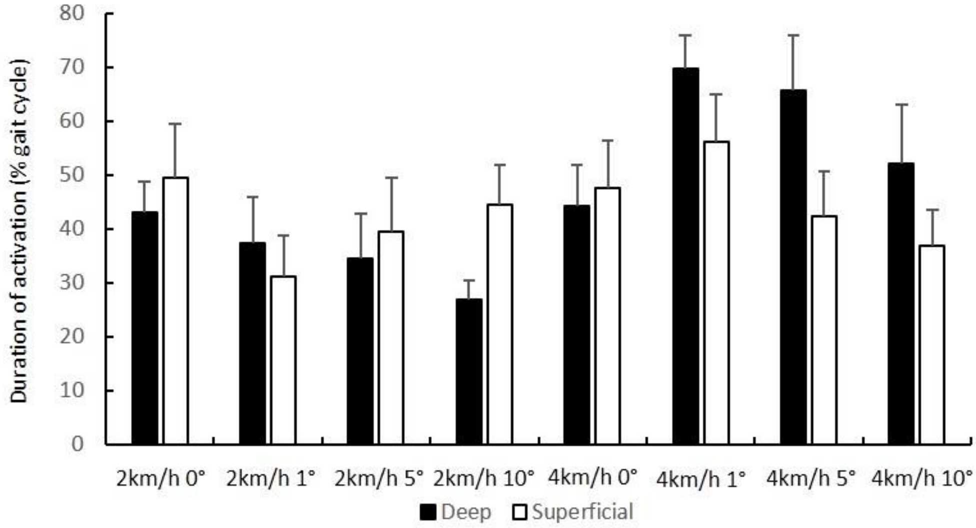
Table 1: Mean (±SD) of spine, hip, knee and ankle range of motion during walking at different speeds and inclinations

	2km/h				*4km/h*			
	0°	1°	5°	10°	0°	1°	5°	10°
Trunk inclination (°)	2.9±0.8	2.8±1.2	2.7±0.6	2.9±1.0	2.6±0.6	2.7±0.6	2.6±0.6	2.7±0.8
Lumbar lordosis (°)	4.4±3.8	4.6±3.5	4.2±2.4	4.2±2.9	4.6±2.5	5.3±3.3	5.5±3.4	5.4±3.7
Pelvic rotation (°)	7.6±1.8	7.8±1.5	8.5±2.5	9.6±2.6	14.2±3.4	14.2±2.8	14.2±2.6	14.9±4.0
Hip (°)	26.3±5.5	27.0±3.9	30.0±7.0	33.1±5.9	30.7±6.7	31.9±7.7	34.4±7.1	40.9±7.8
Knee (°)	47.3±4.1	46.3±3.9	46.6±3.7	46.7±3.6	54.6±4.3	54.2±3.6	54.2±3.6	52.9±4.9
Ankle (°)	13.2±3.5	14.0±3.2	15.6±3.7	18.2±4.8	21.2±5.7	22.9±5.9	24.9±6.1	29.9±6.9


Normalized peak EMG amplitude (%)
300
250
200
150
100
50
0
2km/h 1°
2km/h 5°
2km/h 10°
4km/h 0°
4km/h 1°
4km/h 5°
4km/h 10°
Deep
Superficial



Percent of gait cycle for peak EMG amplitude (%)
70
60
50
40
30
20
10
0
2km/h 0°
2km/h 1°
2km/h 5°
2km/h 10°
4km/h 0°
4km/h 1°
4km/h 5°
4km/h 10°
Deep
Superficial


Duration of activation (% gait cycle)
80
70
60
50
40
30
20
10
0
2km/h 0°
2km/h 1°
2km/h 5°
2km/h 10°
4km/h 0°
4km/h 1°
4km/h 5°
4km/h 10°
Deep
Superficial

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